Predicting the effects of a distal femoral cartilage defect using an accurate three-dimensional dynamic finite element model

Lin-Song Ji¹, Yan-Lin Li³, Yan-Ping Wang², Chun-Shan Luo¹, Lu Tong³

Departments of ¹Spine Surgery, ²Gastroenterology, Guizhou Province Orthopedic Hospital, Guiyang, China; ³Department of Sports Medicine, The First Affiliated Hospital of Kunming Medical University, Kunming, China

Received June 30, 2020; Accepted November 2, 2020; Epub February 15, 2021; Published February 28, 2021

Abstract: Objectives: The present study aimed to investigate how variables associated with an osteochondral femur defect affect tibiofemoral contact using three-dimensional (3D) finite element analysis. Methods: We reconstructed a 3D finite element model of a human knee joint, including the bone structure, cartilage, meniscus and tendons. Then, we built an 8-mm-diameter defect on the weight-bearing area of femoral condyle cartilage in finite element mode. At various flexion angles ranging from 0° to 90°, a 350-N vertical tibial plateau load and a 400-N traction force in the quadriceps direction were applied to the unconstrained proximal femur. After constructing the 3D model, Abaqus software was used for finite element analyses and calculations to examine the effect of an 8-mm defect in the medial femoral condyle (high-weight-bearing region) on the stress of the cartilage and meniscus at 0°, 30°, 60°, and 90° of knee flexion. Results: The results indicated that the forces in the tibiofemoral knee joint increased obviously at 0°, 30°, and 60° of knee flexion during compressive loading. However, at a flexion angle of 90°, the difference in peak stress was not statistically significant. Conclusion: This study demonstrated a significant increase in compressive stresses in the femoral cartilage, meniscus, and tibial cartilage with an 8-mm defect in the medial femoral condyle at 0°, 30°, 60°, and 90° of knee flexion. The present findings provide a better understanding of degenerative arthritis.

Keywords: Articular cartilage, biomechanics, finite element analysis, knee joint

Introduction

Focal chondral or osteochondral defects present several medical and personal complications; these defects can be painful and disabling and may contribute to the development of degenerative arthritis [1, 2]. Moreover, these defects have a markedly poor capacity for self-repair or even medically assisted repair and are the most common forms of trauma among young and active patients [3]. Clearly, better preventive and therapeutic techniques should be developed, but a lack of understanding of knee joint biomechanics impedes the development of these techniques [4, 5]. Despite several investigations, the specific mechanical behaviour of the knee joint and reasons for injuries remain unclear. To date, many experiments have demonstrated that finite element models (FEMs) can provide profound insights into the mechanical properties of biological tissues, overcoming the recurring obstacles of high time and cost requirements. An effectively developed FEM serves as an incredible asset to anticipate the impacts of various parameters involved, yielding information that is difficult to obtain from tests [6].

We aimed to use an accurate three-dimensional (3D) model of a healthy human knee joint including the patellar tendon (PT), anterior cruciate ligament (ACL), posterior cruciate ligament (PCL), medial collateral ligament (MCL), lateral collateral ligament (LCL), menisci, and articular cartilages. This model can demonstrate articular cartilage and meniscus pressure distributions in locations similar to cartilage defects on imaging studies. The primary
Predicting the effect of femoral cartilage using a model

Table 1. Material mechanical properties of the femur, tibia, cartilage, and meniscus

<table>
<thead>
<tr>
<th>Material</th>
<th>Elastic modulus (MPa)</th>
<th>Poisson’s ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td>Femur</td>
<td>3883.4</td>
<td>0.3</td>
</tr>
<tr>
<td>Tibia</td>
<td>4184.6</td>
<td>0.3</td>
</tr>
<tr>
<td>Cartilage</td>
<td>20</td>
<td>0.46</td>
</tr>
<tr>
<td>Meniscus</td>
<td>59</td>
<td>0.49</td>
</tr>
<tr>
<td>MCL</td>
<td>467</td>
<td>0.46</td>
</tr>
<tr>
<td>LCL</td>
<td>467</td>
<td>0.46</td>
</tr>
</tbody>
</table>

Table 2. Material parameters of ligaments

<table>
<thead>
<tr>
<th>Ligament</th>
<th>C₁</th>
<th>D</th>
</tr>
</thead>
<tbody>
<tr>
<td>MCL</td>
<td>1.44</td>
<td>0.00126</td>
</tr>
<tr>
<td>LCL</td>
<td>1.44</td>
<td>0.00126</td>
</tr>
<tr>
<td>ACL</td>
<td>1.95</td>
<td>0.00683</td>
</tr>
<tr>
<td>PCL</td>
<td>3.25</td>
<td>0.0041</td>
</tr>
<tr>
<td>PT</td>
<td>2.75</td>
<td>0.00484</td>
</tr>
</tbody>
</table>

The goal of developing this model was to determine the extent to which an osteochondral defect in the weight-bearing region of the femoral condyle influences the contact conduct of the knee joint during compressive stress loading. The findings may offer some insights into advancing cartilage defect repair.

Material and methods

Three-dimensional model and material properties

An accurate three-dimensional (3D) model of the left knee of a 25-year-old healthy male (height, 1.78 m; weight, 70 kg) was reconstructed based on magnetic resonance imaging (MRI) data. The following associated grouping and parameters were utilized: T1-weighted fat-suppressed gradient recall; repetition time, 11,000 ms; echo time, 25 ms; matrix size, 192/320; and slice thickness, 1.0 mm. MRI images were processed using Mimics 14.11 (Materialise, Leuven, Belgium) and Geomagic Studio 2012 (MSC Corporation, NC, USA) to construct the accurate three-dimensional model [7]. The model was then imported and meshed in Abaqus 6.9 (Simulia, RI, USA).

The model developed in this study incorporated data from previous reports defining bone, cartilage, meniscus, and ligaments as material properties [8]. Femoral cartilage, menisci, and tibial cartilage were presumed to act as linear elastic materials [8]. The elastic modulus for the incompressible matrix of menisci was 59 MPa with a Poisson’s ratio of 0.49 [9]. The elastic modulus for the cartilage was 20 MPa with a Poisson’s ratio of 0.46 as reported in previous joint model studies [10]. All ligaments were assumed to function as hyperelastic material. The constitutive relations model of the material was the neo-Hookean model (Tables 1 and 2).

The neo-Hookean function was evaluated as follows [11]:

\[
\psi = \frac{1}{2D} \ln(J^2) + C_1 (\frac{J_0}{J} - 3) + F_2 (\lambda)
\]

where \(C_1\) is a neo-Hookean constant, and \(D\) is the bulk modulus.

Each tissue of the knee joint was meshed individually. The cartilage, ligaments, and menisci were all divided by tetrahedral solid units. The grid size was set to 1 mm, and the unit was set to a full integration unit. To allow comparisons of the clinical results of the present model with the findings of Yang [12] and Darryl [13], we modelled the smooth, frictionless contact surfaces between the femoral condyle cartilage and the tibial plateau cartilage, the femur and the meniscus articular surface, and the tibial and meniscus articular surface. Given the large elastic modulus of the subchondral bone, it had no significant impact on the calculation and analysis. The subchondral bone was not clearly defined, and the subchondral bone was regarded as a rigid body.

Loads

According to arthroscopic studies, approximately 60% of focal defects are located in the femoral cartilage, and more than half of focal defects in the femoral cartilage occur in the medial femoral condyle [14]. Thus, we created a model of an 8-mm defect in the medial femoral condyle. After constructing the 3D model, Abaqus software was used for finite element analysis and calculations to examine the effect of an 8-mm defect in the medial femoral condyle (high-weight-bearing regions) on the compressive stresses in the articular cartilage and meniscus at 0°, 30°, 60°, and 90° of knee flexion. A 350-N vertical tibial plateau load and a 400-N traction force in the quadriceps direc-
Predicting the effect of femoral cartilage using a model

In this analysis, the distal tibia and fibula were completely restrained and fixed, while the femur was completely free without any constraints. A Von Mises equivalent stress nephogram was used to predict the peak stress of the articular cartilage and meniscus.

Statistical analysis

Data were recorded as the mean ± SD and analysed using SPSS 22.0 (SPSS, IL, USA). A t-test with two independent samples was performed. (P < 0.05) indicates a significant difference.

Results

The model of the knee joint retained its significant individual features. The model with bone and ligament was very similar to the exact anatomy and mirrored their separate anatomical shapes (Figure 1).

When applying a 400-N quadriceps load and a 350-N axial load on the distal femur, peak stresses were found in each part of the cartilage and meniscus. As shown in (Figures 2-5), the maximum equivalent stress was significant-

ly different between the medial condyle cartilage and the lateral condyle of the femur cartilage at 0°-90° of knee flexion (P < 0.05). However, the maximum equivalent stress of the lateral femoral condyle cartilage was lower than that obtained from the medial femoral condyle cartilage. The maximum equivalent stress statistically differed between the medial tibial cartilage and the lateral tibial cartilage at 0°-90° of knee flexion (P < 0.05). However, the maximum equivalent stress of the lateral tibial cartilage was lower than that obtained from the medial tibial cartilage. Similar results were obtained from the lateral meniscus compared with the stress of the medial meniscus.

As shown in (Figures 6-9), the maximum equivalent stress statistically differed between the medial femoral condyle cartilage and the lateral femoral condyle cartilage at 0°-90° of knee flexion. Significant differences in the femoral medial and lateral condyle cartilage, medial and lateral tibial plateau cartilage, and medial and lateral meniscus stress were noted between the two groups at a 0° angle of knee flexion (P < 0.05) (Tables 3 and 4); the increases were 64.8%, 70.9%, 83.6%, 140.9% and 24.7%, respectively (Figure 9 and Table 5).

A statistically significant difference in the value of articular cartilage and meniscus stress was noted between the cartilage defect and the normal group at 30° knee flexion (P < 0.05) (Figure 10 and Table 4).

A statistically significant difference in the value of articular cartilage and meniscus stress was noted between the cartilage defect and the normal group at 60° knee flexion (P < 0.05) (Figure 11 and Table 4).

No significant difference in the peak strain values of the meniscus was noted in the cartilage defect group at 90° of knee flexion compared with the normal group (P > 0.05) (Figure 11).

Discussion

Exploring compressive stress and strain has been a general topic of interest in clinical...
Predicting the effect of femoral cartilage using a model


research. However, such experimental clinical studies can be quite complex largely because the thickness and strength of the articular cartilage are different in different parts. An 8-mm defect was created in the medial femoral condyle because arthroscopy usually detects damage in this area [16-18]. Arthroscopy is currently the primary means of diagnosing and treating knee disorders.

The stability of the human knee is provided by articular cartilage, muscles, ligaments and...
menisci. The structures should be reconstructed in the 3D knee joint model when performing this analysis due to their importance [19, 20]. Most of the previous 3D models did not reconstruct the surrounding ligaments and cartilage [21, 22], which led to the loss of ligament constraints on the knee during finite element analysis. Moreover, Halonen et al emphasized the importance of patella and quadriceps forces in a study of knee cartilage during a gait cycle.
The present knee FEM was composed of 14 structures, including 5 main ligaments (the PT, ACL, PCL, MCL, and LCL). No studies have demonstrated that different joint flexion angles modify the contact variables or have investigated biomechanical changes in the knee. Peña et al. [9], Marchi et al. [24] and Dong et al. [25] analysed only the static conditions of the knee at 0° flexion. Schinhan et al showed that a 7-mm cartilage defect caused an overload of ventral stress.
Predicting the effect of femoral cartilage using a model

Predicting the effect of femoral cartilage using a model

the corresponding site, resulting in more severe degenerative changes in the cartilage compared with a 14-mm cartilage defect [26]. Thus, an 8-mm local full-thickness cartilage defect at the high-weight-bearing areas of the medial femoral condyle surface was selected for lesion simulations. The study sought to analyse contact variables in the articular cartilage and meniscus at 0°, 30°, 60°, and 90° of knee flexion.

The experiments in the present study simulated biomechanical changes resulting from femoral condyle cartilage defects, especially on the adjacent cartilage defects and menisci. The results obtained from Abaqus finite element analysis software indicated that the medial femoral condyle cartilage defect generated an abnormally increasing pressure surrounding the cartilage defect of the medial femoral condyle accompanied by an increase in lateral femoral condyle cartilage secondary stress. Similar results were also obtained by Shelbourne et al. [27]. Finite element analysis results indicate that the peak stress-strain zones of the femoral condyle cartilage appeared in high-weight-bearing areas at different body flexion angles. This effect was similar to that found in previous clinical studies, where the incidence of medial condyle cartilage and lateral condyle cartilage degeneration gradually increased after the medial femoral condyle cartilage was injured [9, 25, 26, 28].

The compressive stresses in the femoral cartilage, meniscus, and tibial cartilage significantly increased at 0°-30°-60° of knee flexion. The cartilage defect group at 90° of knee flexion showed no significant differences in the peak values of the meniscus compared with the normal group. Similar results were obtained by Darryl et al. and Guettler et al. [13, 29]. Darryl investigated only contact variables in the tibiofemoral joint at 0° flexion, and the location of the osteochondral defect was different from that in the present study [13]. Guettler also found that the stress in the bilateral femoral condyle and bilateral tibia plateau increased significantly at 30° of knee flexion. The maximal compressive stress in the medial compartment was higher than that in the lateral compartment in healthy groups. Similar results were obtained by Yang et al. and Carter et al. [12, 30].

Typically, the non-weight-bearing area of the tibiofemoral articular surface and the normal fluid pressure are important factors in maintaining normal articular cartilage nutrition [31]. The high contact pressures adjacent to the defect may interfere with the ability of cartilage around the defect to function normally [29, 32]. Additionally, a change in the weight-bearing area of the tibiofemoral joint can cause intraarticular synovial fluid flow barriers, leaving the affected cartilage with fewer nutrients from the synovial fluid and creating difficulties for cartilage cells in obtaining sufficient water and nutrients to repair cartilage defects [31, 32]. Cristiani et al. [33] and Balazs et al. [34] confirmed that a change in the stress in the cartilage and meniscus is one of the most important factors causing cartilage degeneration and meniscus injury.

In summary, the present study investigated biomechanical changes caused by common

![Figure 9](image)
Predicting the effect of femoral cartilage using a model

Table 3. The maximum Von Mises stress at the cartilage of the femoral condyle, tibial plateau and meniscus at 0°, 30°, 60°, 90° flexion angles (MPa)

<table>
<thead>
<tr>
<th></th>
<th>0°</th>
<th>30°</th>
<th>60°</th>
<th>90°</th>
</tr>
</thead>
<tbody>
<tr>
<td>The medial femoral condyle cartilage</td>
<td>3.47±0.12</td>
<td>3.64±0.09</td>
<td>2.74±0.05</td>
<td>2.95±0.16</td>
</tr>
<tr>
<td>The lateral femoral condyle cartilage</td>
<td>2.51±0.07</td>
<td>2.21±0.02</td>
<td>2.60±0.01</td>
<td>1.96±0.15</td>
</tr>
<tr>
<td>The medial tibial plateau cartilage</td>
<td>6.27±0.11</td>
<td>7.41±0.05</td>
<td>7.05±0.18</td>
<td>5.67±0.22</td>
</tr>
<tr>
<td>The lateral tibial plateau cartilage</td>
<td>4.70±0.15</td>
<td>5.56±0.22</td>
<td>5.29±0.24</td>
<td>4.25±0.23</td>
</tr>
<tr>
<td>The medial meniscus</td>
<td>4.59±0.08</td>
<td>6.52±0.09</td>
<td>5.04±0.07</td>
<td>6.19±0.25</td>
</tr>
<tr>
<td>The lateral meniscus</td>
<td>6.71±0.10</td>
<td>7.13±0.01</td>
<td>6.05±0.29</td>
<td>7.44±0.16</td>
</tr>
</tbody>
</table>

Table 4. Results of the statistical analysis between the normal and defective groups

<table>
<thead>
<tr>
<th></th>
<th>0°</th>
<th>30°</th>
<th>60°</th>
<th>90°</th>
</tr>
</thead>
<tbody>
<tr>
<td>The medial femoral condyle cartilage</td>
<td>&lt; 0.001*</td>
<td>&lt; 0.001*</td>
<td>&lt; 0.001*</td>
<td>0.07</td>
</tr>
<tr>
<td>The lateral femoral condyle cartilage</td>
<td>&lt; 0.001*</td>
<td>&lt; 0.001*</td>
<td>&lt; 0.001*</td>
<td>0.09</td>
</tr>
<tr>
<td>The medial tibial plateau cartilage</td>
<td>&lt; 0.001*</td>
<td>&lt; 0.001*</td>
<td>&lt; 0.001*</td>
<td>0.13</td>
</tr>
<tr>
<td>The lateral tibial plateau cartilage</td>
<td>&lt; 0.001*</td>
<td>&lt; 0.001*</td>
<td>&lt; 0.001*</td>
<td>0.21</td>
</tr>
<tr>
<td>The medial meniscus</td>
<td>&lt; 0.001*</td>
<td>&lt; 0.001*</td>
<td>&lt; 0.001*</td>
<td>0.24</td>
</tr>
<tr>
<td>The lateral meniscus</td>
<td>&lt; 0.001*</td>
<td>0.001*</td>
<td>&lt; 0.001*</td>
<td>0.058</td>
</tr>
</tbody>
</table>

*Statistically significant (P < 0.05).

Table 5. Results of the percentage increments in the normal group with respect to the defective group

<table>
<thead>
<tr>
<th></th>
<th>0°</th>
<th>30°</th>
<th>60°</th>
</tr>
</thead>
<tbody>
<tr>
<td>The medial femoral condyle cartilage</td>
<td>64.8%</td>
<td>75.8%</td>
<td>90.9%</td>
</tr>
<tr>
<td>The lateral femoral condyle cartilage</td>
<td>70.9%</td>
<td>95.9%</td>
<td>67.6%</td>
</tr>
<tr>
<td>The medial tibial plateau cartilage</td>
<td>83.6%</td>
<td>58.6%</td>
<td>76.0%</td>
</tr>
<tr>
<td>The lateral tibial plateau cartilage</td>
<td>83.6%</td>
<td>58%</td>
<td>75.9%</td>
</tr>
<tr>
<td>The medial meniscus</td>
<td>140.9%</td>
<td>72.1%</td>
<td>109.5%</td>
</tr>
<tr>
<td>The lateral meniscus</td>
<td>24.7%</td>
<td>30.5%</td>
<td>16.3%</td>
</tr>
</tbody>
</table>

Figure 10. Histogram of the peak stress force on the articular cartilage and meniscus in the normal and cartilage defect groups at a 30° flexion angle. Asterisks indicate values that were significantly different between the groups, *P < 0.001.

Conclusion

This study demonstrated that with an 8-mm defect in the osteochondral defects at 0°, 30°, 60°, and 90° of knee flexion in the presence of physiological compression forces. However, the study had several limitations. First, the model was simulated for only one size; hence, the position of osteochondral defects in degenerative joint pain was a limitation. Second, the cartilage defect had a round shape, which appears to be non-physiological in the clinical setting. Although the comparison cannot be considered a perfect model, these findings will aid in a better understanding of degenerative arthritis.
Predicting the effect of femoral cartilage using a model

medial femoral condyle at 0°, 30°, 60°, and 90° of knee flexion, the compressive stresses in the femoral cartilage, meniscus, and tibial cartilage significantly increased. Changes in stress distributions are of great significance in long-term degeneration of adjacent cartilage defects.

Acknowledgements

Innovation Team Project in Yunnan Province (No. 2014HC018) and National Cooperation Foundation of Yunnan (No. 2013IA004).

Disclosure of conflict of interest

None.

Address correspondence to: Yan-Lin Li, Department of Sports Medicine, The First Affiliated Hospital of Kunming Medical University, Kunming, China. E-mail: 852387873@qq.com

References


Predicting the effect of femoral cartilage using a model


